INTRODUCTION
Every year, approximately 80,000 to 150,000 ACL tears occur.[1,2] Post-injury, ACL reconstruction (ACLR) is often recommended to restore functional stability and prevent long-term joint degradation. However, while surgical techniques have improved, individuals with ACLR have been shown to have a higher risk of developing osteoarthritis (OA) and a higher rate of re-injury.[3] The higher incidence and earlier onset of knee OA in individuals who have undergone ACLR may be a result of a post-injury movement strategy that utilizes a higher degree of muscle co-contraction of the hamstrings and the quadriceps muscles during landing to increase joint stability.[4] This increased co-contraction leads to a “stiff” landing pattern, decreases shock absorption, and increases the ground reaction forces. These changes in lower extremity biomechanics also may result in higher forces which likely place the knee cartilage under greater stress and higher risk of OA development and re-injury.

The objective of this pilot study was to develop a finite element (FE) model that incorporates subject specific muscle force data and knee joint kinematics to compare the stress in knee cartilage in young females who have undergone ACLR with healthy controls. 3-D FE knee models were created to estimate the normal compressive stress at the tibiofemoral and patellofemoral joint during a drop-landing task.

METHODS
One subject post-ACLR and an age, activity and size matched healthy control subject were used to determine the knee joint forces and cartilage stresses during a drop landing task. The FE model created for this preliminary investigation consisted of both the tibiofemoral and patellofemoral joint and was constructed from sagittal view MRI of the knee using 3-D solid modeling software. A quasi-static simulation was performed with the knee model at 60° of knee flexion with the joint reaction forces (JRF) calculated with inverse dynamics equations and muscle forces calculated from a subject specific electromyography (EMG)-driven muscle force model, Fig. 1.

Figure 1 (A) Anterior view and (B) posterior view of a solid model of the left knee. The location of the muscles and patella tendon were determined from the MRI.

Axial and sagittal images of each subject’s dominant leg were obtained using a 3.0 Tesla MRI system (GE Signa HDx 3.0 T) using a spin-echo pulse sequence (USC Imaging Center, USC Healthcare Consultation Center II).

3-D motion analysis was performed using an 8 camera motion analysis system (Vicon 612; Oxford Metrics, Oxford, UK) at a rate of 250 Hz. Ground reaction forces (GRF) were collected at a rate of 1500 Hz using 2 AMTI force plates (AMTI, Newton, MA, USA). Reflective markers (14 mm spheres) placed at specific anatomical landmarks were used to determine sagittal plane motion of the pelvis, hip, knee, and ankle. EMG data was recorded at 1500 Hz using preamplified bipolar surface electrodes with the Motion Lab Systems MA-300 EMG recording system (Motion Lab Systems, Baton Rouge, LA).
For the drop-landing task, subjects started from a standing position on a 14 inch high platform in front of 2 force plates. Subjects were instructed to drop onto the force plates (each foot on one plate) and then jump up as high as possible.

The subject-specific EMG-driven model required the following input variables: (1) Lower extremity kinematics and GRFs, (2) EMG: (3) Muscle PCSA of the quadriceps, hamstring, and gastrocnemius muscles (4) Lever arm of the quadriceps, medial and lateral hamstring, and medial and lateral gastrocnemius muscle tendon (5) Patellar tendon orientation. A subject-specific, EMG-driven knee joint model was created for each subject.[5] SIMM software (MusculoGraphics Inc., Chicago, IL) was used to create an anatomical knee joint model. The SIMM model included 10 musculotendon actuators which included information of peak isometric muscle force, optimal muscle-fiber length, pennation angle, and tendon slack length, and was represented as a series of 3-D vectors that were constrained to wrap over underlying structures. To create a subject-specific model, muscle physiological cross-sectional area (PCSA) and lever arm data of the 10 muscles and the patellar tendon orientation measured from MRI were incorporated to the SIMM model for each subject. The subject-specific adjusted muscle force vectors combined with the anthropometry, GRF, and lower extremity defined the total loading conditions at the knee for FE analysis.

Sagittal view MRI were used to create the knee joint geometry.[6] Material models for the soft tissue at the knee were taken from anthropometry, GRF, and lower extremity defined the total loading conditions at the knee for FE analysis.

RESULTS
Inspection of Fig. 2 shows greater stress concentration and greater peak normal compressive stress on the cartilage of the ACLR subject compared to the healthy control subject. The peak normal compressive stress for the ACLR subject on the femoral cartilage was 20.2 and 14.0 MPa on the medial and lateral compartment, respectively and 17.9 and 13.3 MPa on the medial and lateral compartment of the tibia, respectively. The peak normal stress for the control subject at the femoral cartilage was 18.7 and 13.6 MPa on the medial and lateral compartment, respectively and 16.2 and 11.8 MPa on the medial and lateral compartment of the tibia, respectively. The ACLR subject also exhibited greater normal compressive stress compared to the control subject. The maximum normal compressive stress occurred on the lateral portion of the patella cartilage. The peak normal compressive stress for the ACLR subject was 26.8 and 27.6 MPa on the on the femoral cartilage and patella cartilage, respectively. The peak normal compressive stress for the control subject was 16.6 and 22.4 MPa on the femoral cartilage and the patella cartilage, respectively.

DISCUSSION
The results support the hypothesis of greater stress at the knee in subjects post-ACLR may underlie the rapid cartilage changes observed in this population. Although the results of the pilot study appear promising, further investigation is warranted. A larger number of subjects are necessary to derive a final conclusion. The main limitation of the FE models is the ability to validate the cartilage stresses. However, the modeling framework described permits an adequate comparison between groups and within subjects (pre- and post-training).

REFERENCES